ELECTROENCEPHALOGRAPHIC BASELINES IN ASTRONAUT CANDIDATES ESTIMATED BY COMPUTATION AND PATTERN RECOGNITION TECHNIQUES

D.O. Walter, R.T. Kado, J.M. Rhodes and W.R. Adey

Space Biology Laboratory
Brain Research Institute
University of California
Los Angeles, California

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Electrocencephalographic Baselines in Astronaut Candidates
Estimated by Computation and Pattern Recognition Techniques

Methods used in acquisition and analysis of electrophysiological data from 200 astronaut candidates are described. Data from 50 of these subjects have been intensively analyzed in establishment of baselines covering a wide range of states of wakefulness and sleep. Accurately timed physiological stimuli and perceptual and learning tasks were presented to all subjects, thus allowing fine comparison between subjects, and establishment of group means for records from each test situation.

Spectral analyses were performed by digital methods for each of the 18 scalp EEG channels with leads located according to a modified 10-20 plan. From the primary spectral density parameters, averages and variances were calculated for each scalp location for the whole group of 50 subjects. These averages covered resting conditions, recurrent somatic, auditory and visual stimuli, vigilance tasks, and visual discriminations at different levels of difficulty. Similar analyses were performed on 30 subjects in drowsy and sleep states.

In each case, despite wide individual differences between subjects, the group mean and/or pattern of variance in spectral densities for each test condition presented a characteristic pattern. These patterns were consistent with neurophysiological observations on organization of cortico-subcortical interrelations and cerebral systems. Recent evidence relating the scalp EEG to intracellular wave phenomena in cortical neurons is reviewed.

In continuing studies, automated pattern recognition techniques have been applied to the primary outputs of spectral analysis. Preliminary results presented here from 4 subjects indicate an accuracy exceeding 90 per cent in computed assignment of states of wakefulness, based on 10 to 20 second epochs of data from 4 EEG channels, and evaluation of 78 variables. Coherence measurements were of great importance in these studies. The value of EEG records in flight monitoring is reviewed and the feasibility of on-line computation discussed.

With planned extension of manned space flights from days to many weeks, and ultimately to several months in interplanetary flight, physiological monitoring of astronaut status must necessarily cope with new dimensions. Man will assume the role of a mariner, rather than a pilot, and will then be required to meet repair and maintenance schedules based on daily assessment of spacecraft systems. Philosophies of inherent reliability that originated in aircraft development will necessarily be replaced by concepts of modular construction and by unit replacement and repair.

In these circumstances, physiological and psychological assessment based on such parameters as the electrocardiogram and respiration leaves much to be desired in fine evaluation of changes of consciousness from wakefulness to deep sleep. As pointed out elsewhere,[2] such data are even less revealing of transient shifts in alertness and focused attention associated with drowsiness and fatigue. Nor is accuracy of evaluation likely to be adequately incremented by information about eye movement, or verbal response to specific questioning, There is a continuing need for a reliable monitoring system capable of functioning with a passive subject, and even more importantly, on a non-interference basis.[6]

Recent events in manned space flight have emphasized that it should be an uncompromising requirement to collect sufficient physiological data over an appropriately long period from a sufficient series of subjects to allow distinction of those changes representing an immediate reaction to the sudden imposition of weightlessness, to which adaptation may occur in varying degree, from undesirable trends gradually appearing after an initial period of satisfactory physiological status. Direct monitoring of brain electrical activity can reveal both altered diurnal cycling in prolonged flights, as well as any subtle shifts that might accompany brief alerted patterns in the few seconds required for discriminative judgment. Periodicities in sleep-wakefulness cycles are vital to the maintenance of high judgment and performance levels in prolonged flights.

Complexities of multichannel EEG records and differences between subjects have discouraged fine interpretations based on visual examination of paper records. For these reasons, we have performed extensive computer analyses of EEG patterns in data from a population of astronaut candidates. This study has established for the first time that there are common factors that clearly categorize EEG records in such a population for a series of specified behavioral states. The test states have covered a wide gamut from quiet wakefulness to extreme focusing of attention, and the various stages of sleep. Moreover, pattern recognition techniques applied to such data from individual subjects have indicated high reliability in automated assessment of EEG patterns accompanying a spectrum of states of consciousness[28] and have indicated the feasibility of such measures as the basis for a computer based flight monitoring system.

This paper will discuss methods used in the acquisition of a "normative library" of EEG data from 200 astronaut candidates, analysis techniques, and collective and individual patterns detected in 50 of these subjects. The following paper will describe application of these methods to EEG data gathered in space flight. The acquisition of data from the astronaut candidates was performed in the Neurophysiology Laboratory at the Methodist Hospital, Houston, by Dr. P.M. Kellaway and Dr. Maulsby, whose endless patience has assured very high quality records from all 200 subjects. This has minimized problems of artifacts in computation of these records. It is a pleasure to acknowledge their most important contributions to this project.

METHODS

a. Physiological data acquisition system.

Two hundred astronaut candidates were tested in a series of perceptual and learning tasks by means of a programming device.[6,13,29] This device was developed in the UCLA Space Biology Laboratory, and uses a magnetic tape command system to ensure accurate timing from one subject to the next. The tape carries vocal instructions on one track. Command signals to advance projectors and to present somatic, auditory and visual stimuli are carried on multiplexed audio subcarriers, using an IRIG system of narrow-band frequency modulation. In addition to physiological stimuli, each subject was presented with a series of perceptual and learning tasks of increasing difficulty over a period of one hour. Each behavioral situation was designated on the magnetic tape by a 3 figure binary coded decimal number which was recognizable by the computer in subsequent data reduction. All situation and stimulus markers were similarly carried on an IRIG audio subcarrier. Two of these test devices were installed in the Neurophysiology Department at the Methodist Hospital, Houston, for subject testing.

A wide range of physiological parameters were recorded from each subject (Fig. 1), including 18 channels of EEG, electrooculogram (EOG), electrocardiogram (EKG), galvanic skin responses (GSR) and respiration. Scalp leads for the EEG used silver disc electrodes with paraffin fixation. Their placement followed a slightly modified 10-20 schema, hopefully to encourage utilization of this data library by other investigators interested in establishment of EEG patterns related to task performance and behavioral states.

Physiological data was acquired on two once-inch, 14 channel magnetic tapes, one of which carried the majority of the EEG channels, and the other the autonomic measures. Stimulus and situational markers were common to both tape recorders and provided required synchronism between tapes in subsequent calculations of shared parameters, such as coherence. A paper record was also prepared of physiological data, together with BCD situational codes and stimulus markers.

During the test, the subject was comfortably seated in front of the projector screen, with a pushbutton device at his right hand for responses to behavioral tasks. Epochs of physiological stimuli and behavioral tasks were interspersed with "resting" periods in which the subject sat quietly, with eyes open or closed, as specifically commanded. Behavioral tasks ranged from simple auditory vigilance tests, with a requirement to recognize presence or absence of 3 tones in a series presented at 6 second intervals, to recognition of the largest of six circles in a matrix, with a 3 second test epoch in some series, and a 1 second epoch in others. The 3 second task required moderate concentration, whereas the 1 second task was near the limit of performance capability in many subjects.

Additional records were obtained in sleep. Edited segments from light and deep phases and arousal from sleep were analyzed in the same fashion as those from the performing subjects.

b. Computer analysis techniques.

Data tapes were digitized at 200 samples per second, and these digital tapes at 800 b.p.i. were prepared with both physiological signals and event markers. A complete digital "library" has been prepared for all data from

50 of the 200 subjects, and since the computer can readily locate any desired data epoch on command, this digital library has been used in the compilation of results for the group as a whole, and is a source of data in a series of continuing projects.

Computing techniques in neurophysiology have been reviewed elsewhere.[1,4] The need to recognize and characterize patterns in continuous complex processes such as the EEG has led to the use of modern spectral analysis techniques. These have been extensively applied to problems of missile flight and to geophysical data.[8,23] For the electroencephalographer, the power of spectral analysis lies in its ability to reveal the relations between simultaneous brain wave activity in different brain regions, with precise preservation of information about shared frequencies and phase relations, not merely at the dominant frequencies in a spectrum of activities, as in the cross-correlogram, but with equal clarity and precision at each and every frequency in the spectrum.[7,24,25,26,27] The key to application of the digital computer lies in its ability to function as a narrow band filter, with precisely specified characteristics, which can also be modified at will with respect to flat-top, shoulder and skirt.[15,21] These filters have successfully overcome difficulties in design of physical filters with appropriately narrow skirt characteristics, particularly at low frequencies. The digital filter can be considered as having a narrow bandpass, as in an analog filter. A set of digital filters applied to a function of time, such as the EEG, can be considered a version of a Fourier transform.

Since the phase shift of the digital filters used was zero, it was possible to measure phase relations between EEG wave trains at each frequency across the spectrum, as well as shared amplitudes between them at each frequency.

This had led us to routine calculation of the coherence function, as a measure of statistical variability in linear interrelationships between brain regions. As described below, this function is of great value in defining EEG patterns accompanying specified states of sleep and wakefulness. It is expressed as:

$$Coh(f) = MAGS^2(f)/ASX(f)ASY(f)$$

where MAGS(f) is the mean cross-spectral magnitude at frequency f, and ASX(f) is the autospectrum of X and AXY(f) is the autospectrum of Y, at the respective frequencies. Coherence has values between 0 and 1, and is a measure of the linear predictability of activity in any area, on the basis of knowing the activity in any other area; it can also be extended[16] to predictability based on activities in a series of areas.

Initial spectral density calculations were made for each of the 18 EEG leads on the IBM 7094 computer in the UCLA Health Sciences Computing Facility. To synthesize the data, the spectral outputs were averaged for each scalp location over all 50 subjects both in test situations and in selected sleep epochs. A synthetic "head" was thus constructed, with the average for each scalp region presented as a bar graph covering the spectrum from 0 to 25 cycles per second. Further developments of this display technique to provide separate "heads" for different test epochs are presented below. Each hour of subject data required 25 hours of main computation time, wherein multiplications were performed at approximately 500,000 per second. Such a comprehensive initial analysis has allowed selection of variables for a substantially simpler and potentially on-line system.

RESULTS

1. EEG characteristics of a population of 50 subjects over a range of awake states.

After computation of spectra for each subject over all test situations, averages were prepared for all 50 subjects at each scalp location (Fig. 2). This average was over 12 situations (top left), including sitting with eyes closed at rest, eyes open at rest, eyes closed during one per second flash stimuli, and during an auditory vigilance task and visual discriminations at 3 second intervals, and a similar series of more difficult discriminations at 1 second intervals. The contours of these "lumped" spectra were used as the basis for comparison with the spectra for the individual situations. Subsequent "heads" in Fig. 2 show variations about the mean that was established by the average of the 12 situations shown in the top left figure. Spectral densities above the mean at any frequency have bars above the baseline and vice versa; the unit of variation for each bar position was the standard deviation of power at that frequency, among the 12 situations analyzed. It will be noted that spectral densities in the "averaged" head across 12 situations were highest at frequencies below 5 cycles per second in all frontal and central leads, and the alpha peak at 8 to 12 cycles per second in parietal and occipital leads was no larger than this low frequency peak. Careful tests have excluded tape recorder wow and flutter as a source of this low frequency peak. We have noted similar spectral contours in cortical records in the chimpanzee[22] and from depth leads in man.[5] These findings are emphasized since visual inspection of EEG records gives little indication of these high powers at low frequencies.

a. Averages of EEG spectra in testing records and during flashes at 1/sec.

With eyes closed and at rest (Fig. 2, top middle), powers in all frontal leads were below the group average at all frequencies. Central and parietal leads showed a small peak above the mean in the alpha band around 10 cycles/sec. This peak was broader in occipital leads. The power in the remainder of the spectrum from 0 to 25 cycles per second was below the mean by 1 to 2 standard deviations.

These findings contrast sharply with identical analyses during flashes at 1/sec with eyes closed. Although frontal power remained low, there was a marked increase in the power of the alpha peak in central and vertex leads. There was a progressive broadening and increased height in this mid-spectral peak in parietal areas. In the occipital and occipitotemporal region a major increase in spectral power occurred in all frequencies, even though the low flash rate was clearly not responsible for any driving phenomenon.

b. Averages of EEG spectra during auditory vigilance task.

In this task, the subjects were required to press a button each time 3 tones were heard in a succession of tests at intervals of 6 seconds over a period of 8 minutes. Unusual groups were presented at irregular intervals during the task. Eyes were closed throughout.

The resulting EEG averages (Fig. 2, bottom left) produced a pattern distinctly different from that during the slow repetition of flashes, or during visual discriminative performances described below. Vertex, parieto-occipital and temporooccipital leads showed distinctly bimodal spectral distributions, with separate peaks centered around 10 and 20 cycles/sec. In summary, the most striking changes were those in the belt of cortex extending through the temporo-parieto-occipital regions.

c. Averages of EEG spectra during visual discriminative performances.

The task involved identification of minor size differences in sets of 6 circles. The test images were in two series, the first involving successive presentations at 3 sec intervals, and the second at 1 sec intervals.

The concomitant averaged spectral "heads" were clearly different from either resting or vigilance tasks spectra. In the 3 sec tests (Fig. 2, lower middle) there was a sharp increment in frontal activity over the whole spectrum, and this declined gradually towards the central and vertex leads, where the alpha peak noted in the previous test situations was now sharply reduced, and powers were reduced in the alpha band below the multi-situational mean. In the bioccipital lead, powers were reduced below the group mean at all frequencies above 2 cycles/sec.

The trends noted in the 3 sec tests were more obvious in those lasting only 1 sec, but the two "heads" were obviously similar (Fig. 2, bottom right). As indicated below, a clear distinction between EEG patterns in the two cases was possible, however, by utilization of coherence measurements of degrees of shared activity between different cortical regions in the two data sets. Coherence measurements in this study have been described in detail elsewhere.[13]

2. EEG characteristics of a population of 30 subjects in drowsy and sleep states.

Analyses of drowsy and sleep states followed the same procedures as for the awake records. An averaged "head" was prepared for 30 subjects covering states of presleep, sleep and postsleep (Fig. 3A). This was then used as the mean for assessment of spectral variations in the sleep records (Figs. 3B-D and 4 E-H), now using as the unit of variation the standard deviation of the power spectra density at each frequency, among the 7 included situations.

Closing the eyes in preparation for sleep (Fig. 3B) induced a powerful peak in the alpha band that extended further into the frontal region than in resting epochs during task performance. At the same time, a broad but clearly separated peak extending from 15 to 25 cycles occurred in all but the most frontal leads, and increased progressively in power toward the occipital regions. With the onset of drowsiness (Fig. 3C), this picture changed sharply. The alpha peak disappeared completely from all areas, and the high-frequency peak was confined to central and vertex leads, and reduced in amplitude.

Records classified as light sleep (Fig. 3D) include those with "parietal humps" accompanying REM (rapid eye movement) or dream sleep. In view of the superficial resemblance of these records to those in the awake state, it is notable that they are clearly distinguishable in this display from those in task performances (Fig. 2), with a small peak at 4 to 5 cycles/sec in central, parietal and occipital areas, and powers below the mean in the middle range from 5 to 15 cycles/sec, but not higher frequencies. Although the different data bases influences these displays, computed spectral differentiation of this sleep stage with a high reliability is in agreement with studies in the chimpanzee[22] and emphasizes the value of computation over mere visual inspection.

In medium sleep (Fig. 4E), characterized by 14/sec spindles in central and vertex leads, powers at other frequencies remained close to the mean. In deep sleep with high voltage slow waves below 6 cycles/sec (Fig. 4F), powers at higher frequencies was sharply reduced below the mean at frequencies above 15 cycles/sec. Subarousal to an auditory stimulus, inducing "K-complexes" in the EEG records, produced yet another type of spectral contour, which showed highest powers above the mean at the lowest frequencies of the spectrum, falling toward mean values at increasing frequencies around 7 cycles/sec, and showing powers below the mean at frequencies from 7 to 25 cycles/sec (Fig. 4G). Actual arousal (Fig. 4H) was associated with power sharply increased above the mean in lateral frontal areas across most of the spectrum, with gradually declining increments towards the occipital region.

In summary, these EEG patterns accompanying a continuum of behavioral states from strongly focused attention to deep sleep are in accordance with well established data on cerebral organization both in sensory systems, including auditory, visual and somatic pathways, and in associative or integrative mechanisms of the frontal and temporal lobes. The significance of these results does not appear therefore to lie in any casual relationship to the imposed environmental stimuli, but to arise causally in the transaction of information in cerebral systems.

3. Assessment of EEG characteristics in individual subjects; use of pattern recognition techniques.

We have developed display techniques, using two-color methods, that depict for an individual his EEG spectral characteristics in each scalp lead against the contour of the group mean, or against his own mean. However, there is an obvious requirement in medical monitoring to take account of individual variability, by possibly simpler computational techniques, for accurate specification of subject status. For this reason, we have examined the feasibility of automated pattern recognition applied to the output of primary spectral analysis.[28]

Discriminant analysis was applied to the spectral outputs for five situations: eyes closed at rest, eyes open at rest, during the auditory vigilance tasks, and during the two visual discriminative tasks described above. Results will be presented from 4 subjects in 5 situations, with evaluation of 78 variables. A continuing study nearing completion involves 10 subjects in 11 situations to the behavioral situation from which it came. Measurements were derived from four EEG channels: left and right parieto-occipital (P3-01 and Pr-02), vertex (FZ-CZ), and bioccipital (01-02). Each channel's activity was analyzed into four frequency bands (1.5-3.5, 3.5-7.5-12.5, and 12.5-25.0 cycles/sec). For each band, measurements were made of the strength of activity in each channel, mean frequency within the band (the dominant frequency when present), bandwidth within the band (a measure of the regularity of the dominant frequency), and the coherence between pairs of channels.

Details of this discriminative analysis program have been described elsewhere.[28] Briefly, the selection process involves recognition of a single parameter that discriminates EEG segments recorded in different situations, followed by selection of second and subsequent parameters that add most to the discriminating power of the first measurement. The four parameters that best characterized the five situations were: left parieto-occipital alpha band intensity (7.5-12.5 cycles/sec), the mean frequency of theta band activity (3.5-7.5 cycles/sec) in the vertex, the coherence in the theta band between left parietooccipital and vertex, and coherence in the beta band (12.5 to 25.0 cycles /sec) between vertex and bioccipital leads.

Using only the four best measurements for each subject, between 62 and 69 per cent were correctly classified, contrasted with 51 per cent correct for all four subjects as a group. An even greater disparity appeared between group and individual classifications after 15 parameters were selected. Individually, 95, 93, 96 and 90 per cent were correct, whereas for the subjects together, only 65 per cent were correct. Moreover, the distinctions were greatest between those situations most difficult to separate by visual inspection (those associated with two durations of visual discrimination) (Fig. 5). It may therefore be inferred that each individual may have a spatially and numerically characteristic EEG "signature," on the basis of a possibly unique cluster or constellation of EEG parameters that appear with each test situation. Since the analyses involved epochs of only 10 to 20 seconds, from each of the 4 channels, and four parameter types, these findings encourage the possibility of development of a refined and rapid scheme of pilot or astronaut monitoring.

DISCUSSION

Detection of EEG patterning in relation to states of sleep and wakefulness appears to be feasible with high reliability, using computing techniques now well within the state of biomedical engineering arts. It would be unfortunate, however, if these capabilities were viewed as merely offering an empiric evaluation of the essential mechanisms in transaction of information in brain systems. Much has been written in support of the view that the EEG is merely a "noise" in cerebral systems. Only recently has evidence clearly indicated that the EEG recorded grossly from the scalp, or from the brain surface, arises in an intracellular wave process,[9,10,14] and that there is a strong relationship in spectral density distributions between these intracellular waves and the gross EEG, although the coherence between the two is usually low.[3,11,12]

The search for patterns in the complexities of EEG wave trains had their origins in the detection of "evoked potentials," transient electrical events detectable from the vast sea of EEG waves by their time-locked relationship to a preceding equally brief stimulus, such as light flashes, clicks or somatic stimuli. Their detection in brain electrical activity has rested on a variety of averaging techniques. These methods have demonstrated in man and animals an impressive series of changes in evoked potential configurations in response to central and peripheral manipulations. Yet their value in an understanding of cerebral mechanisms has remained limited, since we do not normally live in an environment viewed tachistoscopically by light flashes or similar uncomfortably brief sensory influxes. Not only is our sensory influx a continuum, but the brain likewise is continuously active in the waking state, engaged in "housekeeping" processes, on which the arrival of a sensory volley represents a perturbation.

Early attempts to treat the EEG both as a continuous process and as an essential phenomenon in the transaction of information in cerebral systems were based on frequency analysis,[17,20,30] using analog filtering techniques. Such analyses were immediately beset with an insuperable difficulty. Due to complex and variable phase shifting of the EEG signal in its passage through the filter, no information could be derived about patterning of interrelations between different brain regions on a wave-by-wave basis, and at each frequency across the spectrum. It is in this ability to derive such phase relations in a wealth of essential detail that the digital filter, used in EEG spectral analysis with calculation of coherence, has opened new windows on complex but consistant patterns in the EEG accompanying task performance and in sleep states. Digital analyses of waking and sleeping EEG records by Johnson, Nute, Hustin and Lubin[19] support our findings.

This quality of complexity in the EEG has long been recognized in normal and abnormal subjects,[18] and within and between individuals. The present study appears to have taken successful account of the wide range of individual differences in the specification of collective means or group characteristics, as well as successfully deriving a patterned EEG "signature" for individual subjects. Without due regard for such complexities, there is no reason to anticipate development of adequate or reliable analysis methods, for complexity is surely of the essence in the functioning of brain tissue, wherein our understanding has been unduly delayed by the futile pursuit of analyses and cerebral models based on oversimplified "nerve nets."[3]

On-line analysis of EEG records in classification of behavioral states would appear an important objective for the medical flight monitor, or for pilot warning in anticipation of defective attention through drowsiness, fatigue or problems of environmental support. Hopefully, the present study has established a baseline for use in future flight studies. It has also indicated the possibility of using a small, special purpose computer that would work with data from 3 or 4 EEG channels, and achieve a classification on the basis of evaluation of a small number of variables for each channel, including spectral densities and bandwidths, dominant frequencies and coherence functions. Only a limited amount of data would then be telemetered to a larger flight monitoring computer.[6]

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LEGENDS TO FIGURES

Fig. 1. Scheme for presentation of auditory and visual stimuli and learning situations from a master control magnetic tape to a subject for computer analysis of EEG records. The system utilizes a twin-track recorder for a voice protocol and command signals. The latter are carried on IRIG subcarrier oscillators. The data acquisition tapes include the command signals and the EEG records.

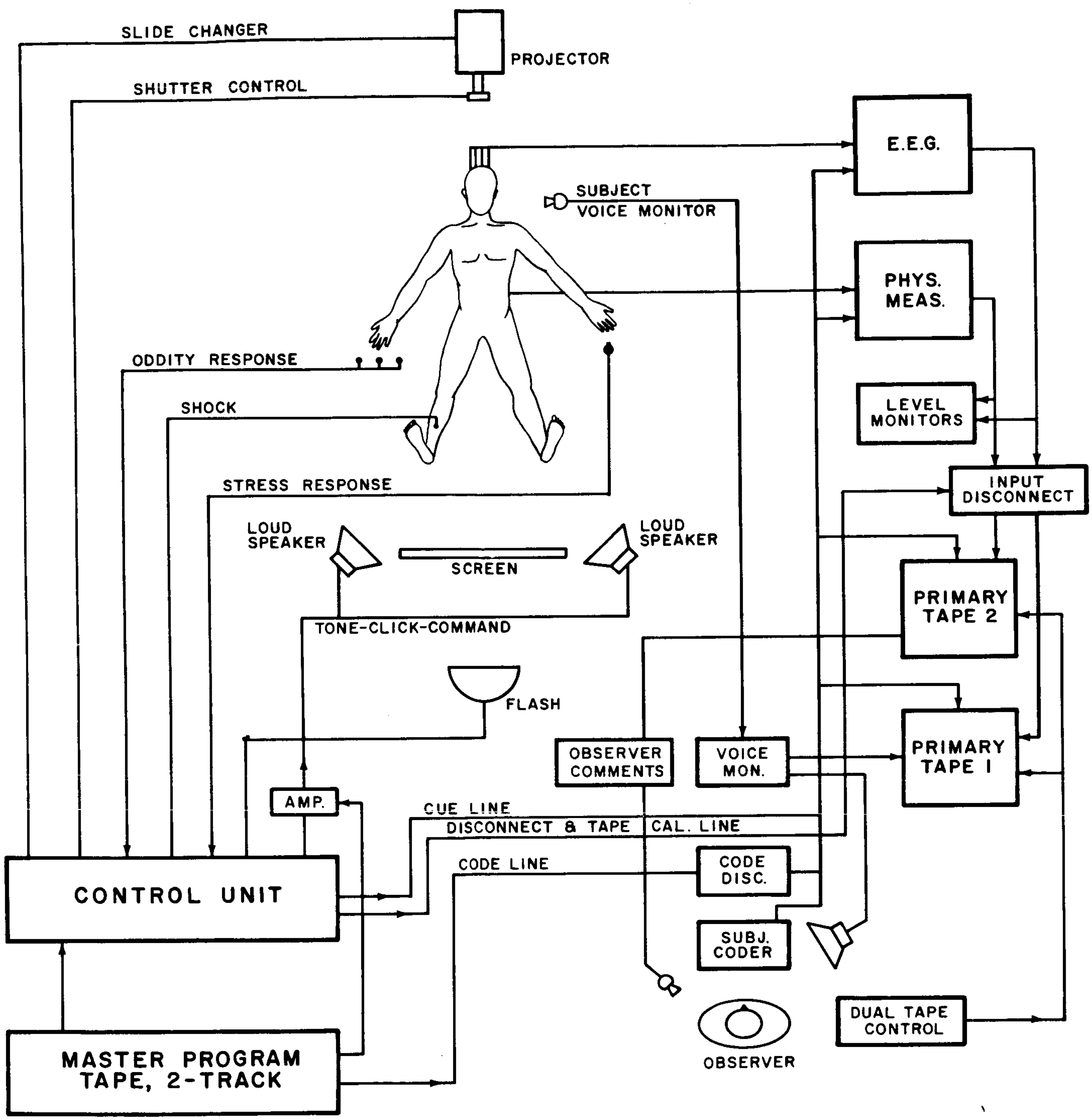
Fig. 2. Averaged spectral densities over the range 0 to 25 cycles per second for a population of 50 subjects, with each spectrum presented as a series of bars at 1 cycle per second intervals, and located at the appropriate location on the scalp. The top left figure is an average for all subjects across 12 situations (see text). The contour of this average was then used as the mean against which to measure deviations in the succeeding 5 situations, with powers at any frequency above the mean shown as a bar above the baseline and vice versa. Calibrations for average over 12 situations in microvolts squared per second per cycle; for the separate situations, in standard deviations. (French, Adey and Walter, 1966).

Fig. 3. Averages prepared as in Fig. 2 for a population of 30 subjects in 7 stages of presleep, sleep and postsleep, and the separate averages, with display of deviation from the mean developed in Fig. 9A, in the succeeding heads for eyes closed awake, drowsy and light sleep records. (French, Adey and Walter, 1966).

Fig. 4. Averages prepared as in Figs. 2 and 3 for medium sleep, deep sleep, subarousal, and arousal to auditory stimuli. (French, Adey and Walter, 1966).

Fig. 5. Pattern recognition techniques applied to spectral outputs from 4 subjects, separately and jointly, with development of a matrix display of automated classifications for 5 situations: EC-R, eyes closed resting; EO-R, eyes open resting; EO-R, eyes open resting; EC-T, eyes closed while performing an auditory vigilance task; EO-T-3, performing moderately difficult visual discriminations in 3 sec; EO-T-1, performing difficult visual discriminations in 1 sec. (Walter, Rhodes and Adey, 1966).

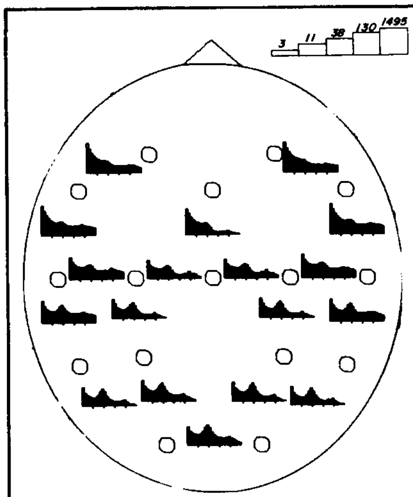
APPENDIX I A



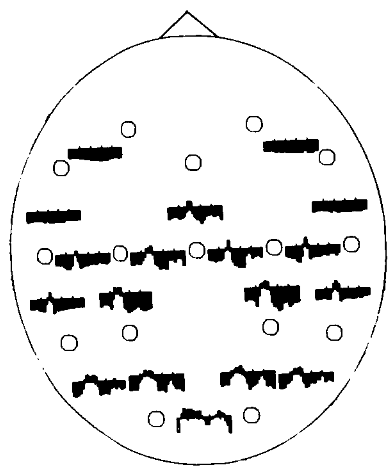
PSYCHO-PHYSIOLOGICAL TESTING AND
DATA ACQUISITION SYSTEM BLOCK DIAGRAM

RESPONSES OF ELECTROENCEPHALOGRAM TO DIFFERING SITUATIONS

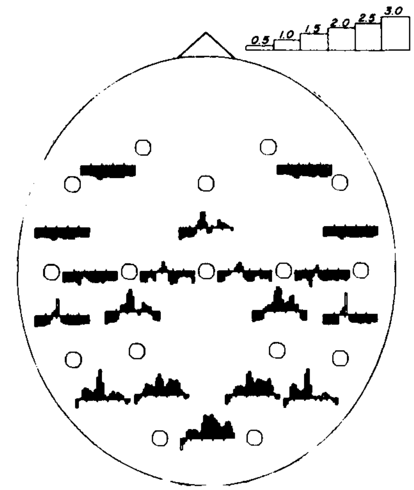
TOPO-SPECTROGRAPHIC VARIATIONS OF
AVERAGES OVER FIFTY ASTRONAUT CANDIDATES



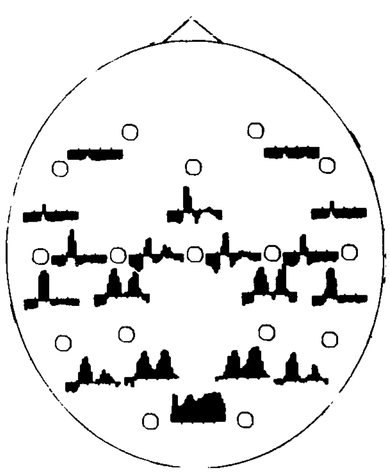
AVERAGE OVER
TWELVE SITUATIONS



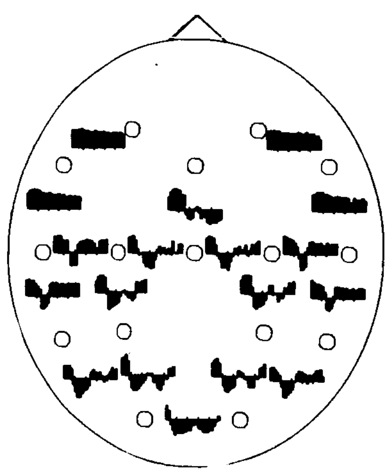
EYES CLOSED
INTERSTIMULUS RESTS



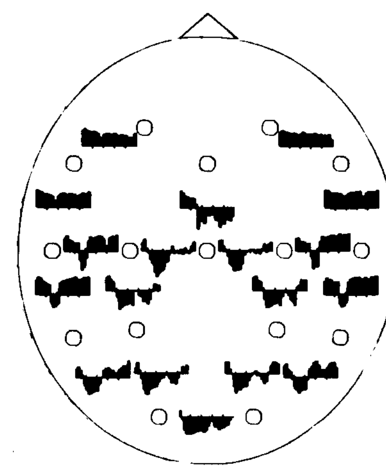
EYES CLOSED
FLASHES, ONE PER SEC.



EYES CLOSED
VIGILANCE TEST
TONES & BUTTON PRESSING



EYES OPEN
VISUAL DISCRIMINATION
3 SEC. EXPOSURE

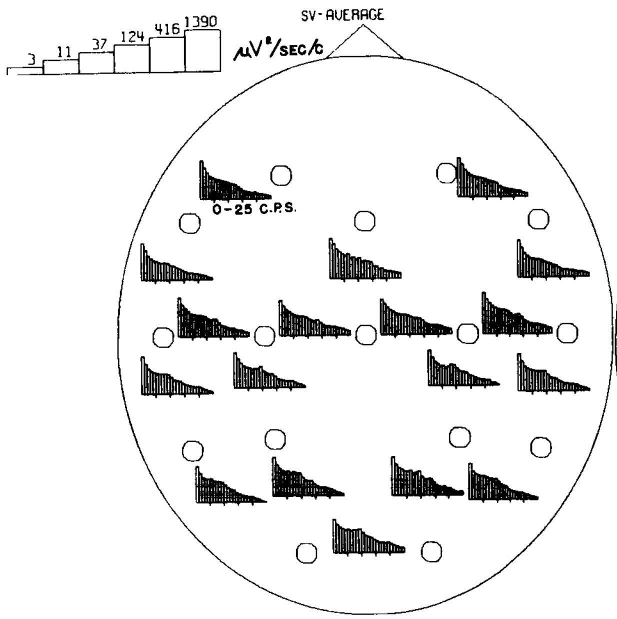


EYES OPEN
VISUAL DISCRIMINATION
I SEC. EXPOSURE

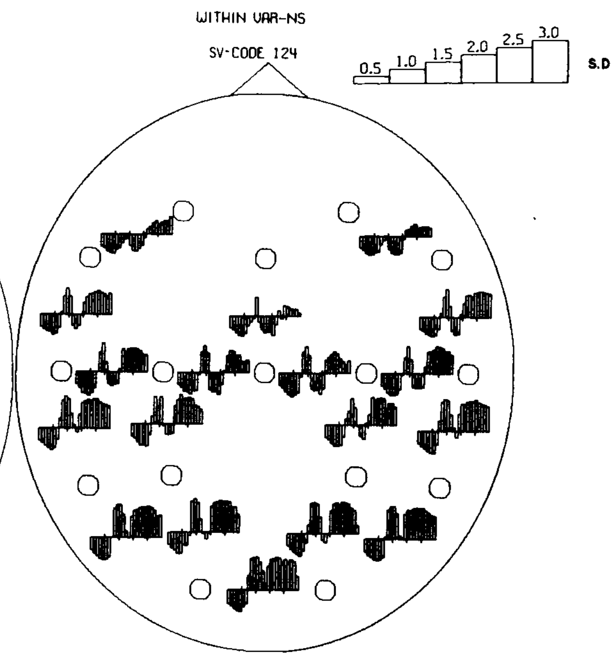
ELECTROENCEPHALOGRAPHIC CHARACTERISTICS OF SLEEP

TOPOSPECTROGRAPHIC VARIATIONS OF AVERAGES OVER 30 ASTRONAUT CANDIDATES

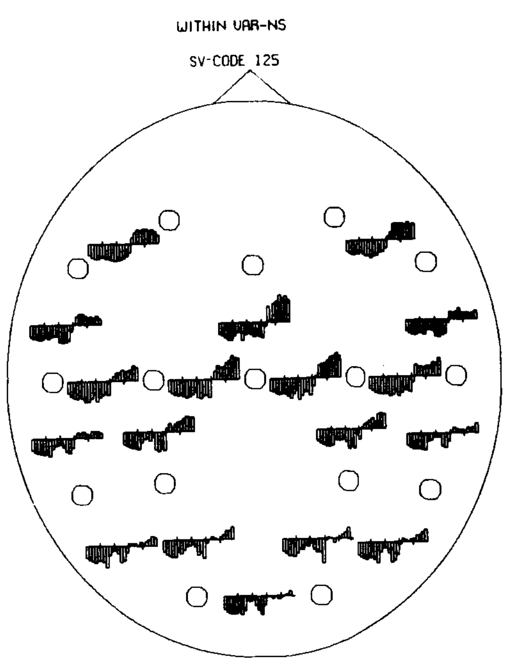
A. AVERAGES OVER 7 STAGES OF PRESLEEP, SLEEP & POSTSLEEP



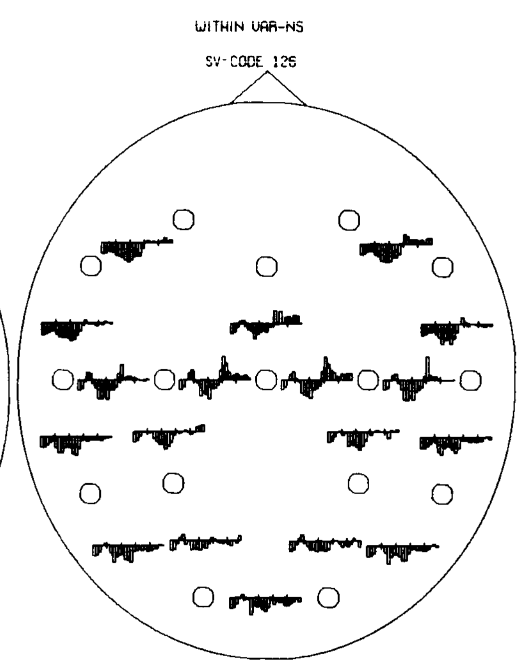
B. SLEEP Ø — EYES CLOSED, AWAKE



C. SLEEP I. "DRIFTING" OR DROWSY



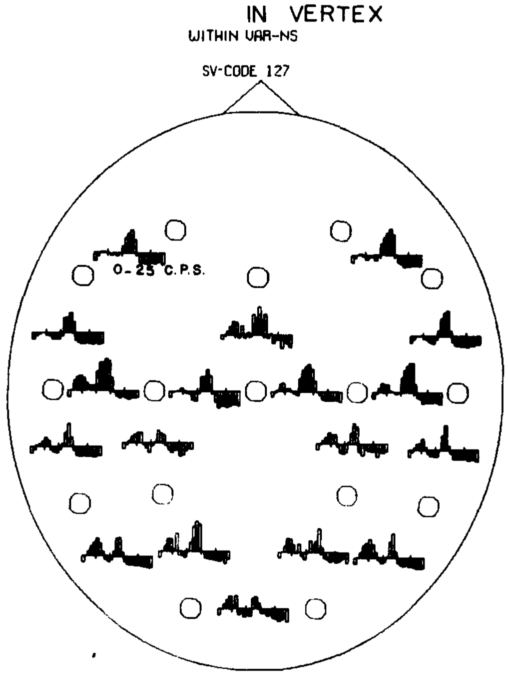
D. SLEEP II. LIGHT SLEEP—"PARIETAL HUMPS"



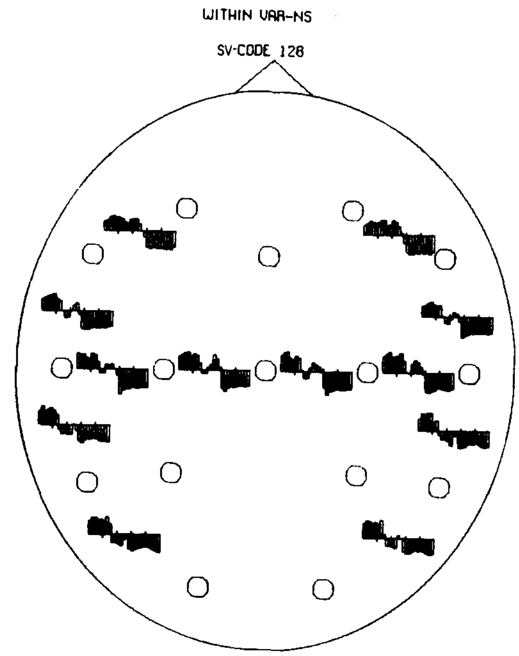
ELECTROENCEPHALOGRAPHIC CHARACTERISTICS OF SLEEP

TOPOSPECTROGRAPHIC VARIATIONS OF AVERAGES OVER 30 ASTRONAUT CANDIDATES

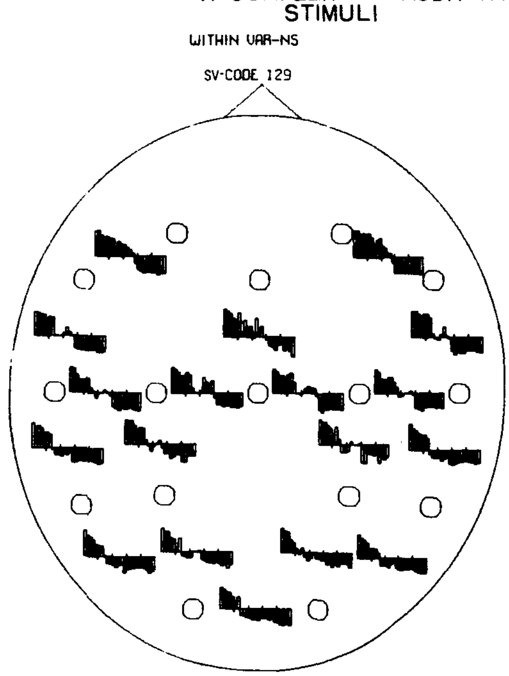
E. MEDIUM SLEEP — 14/SEC SPINDLES IN VERTEX



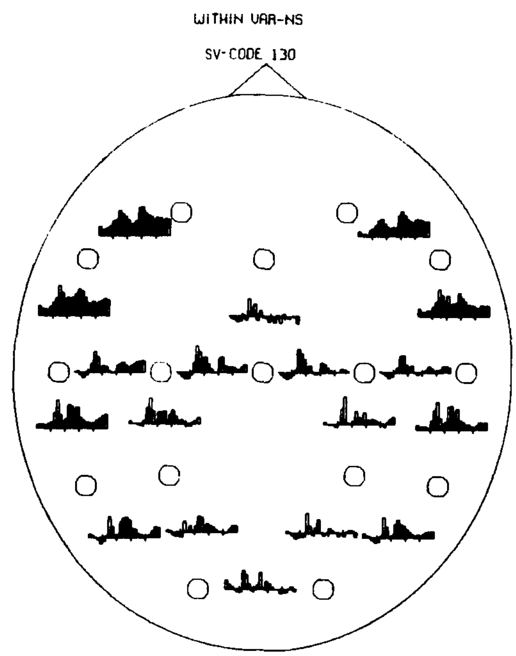
F. DEEP SLEEP — HIGH VOLTAGE SLOW WAVES



G. SUB-AROUSAL — "K-COMPLEX" TO AUDITORY STIMULI



H. AROUSAL TO AUDITORY STIMULI



AUTOMATIC CLASSIFICATION BY BEST
4 MEASUREMENTS